FIG. 1

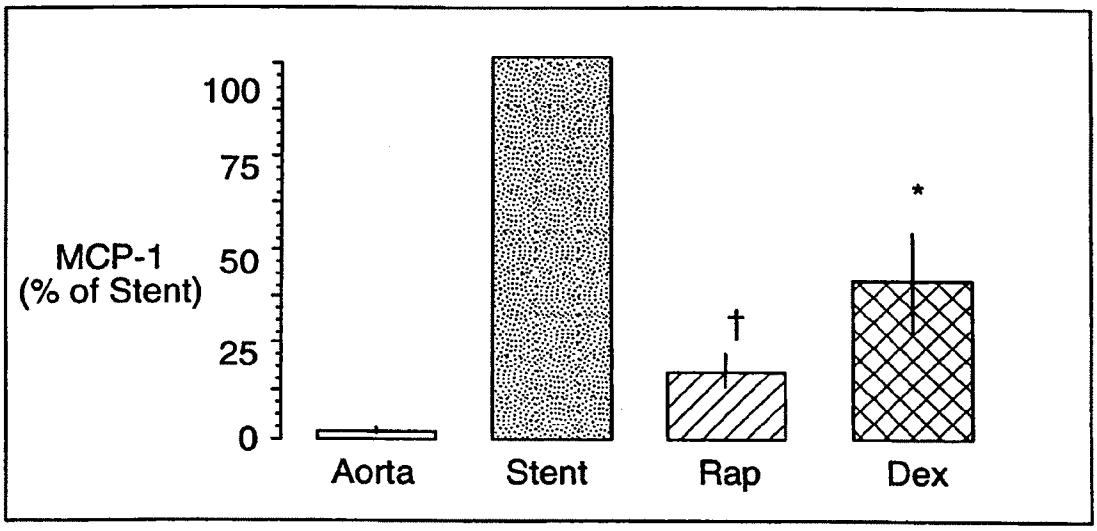


FIG. 2

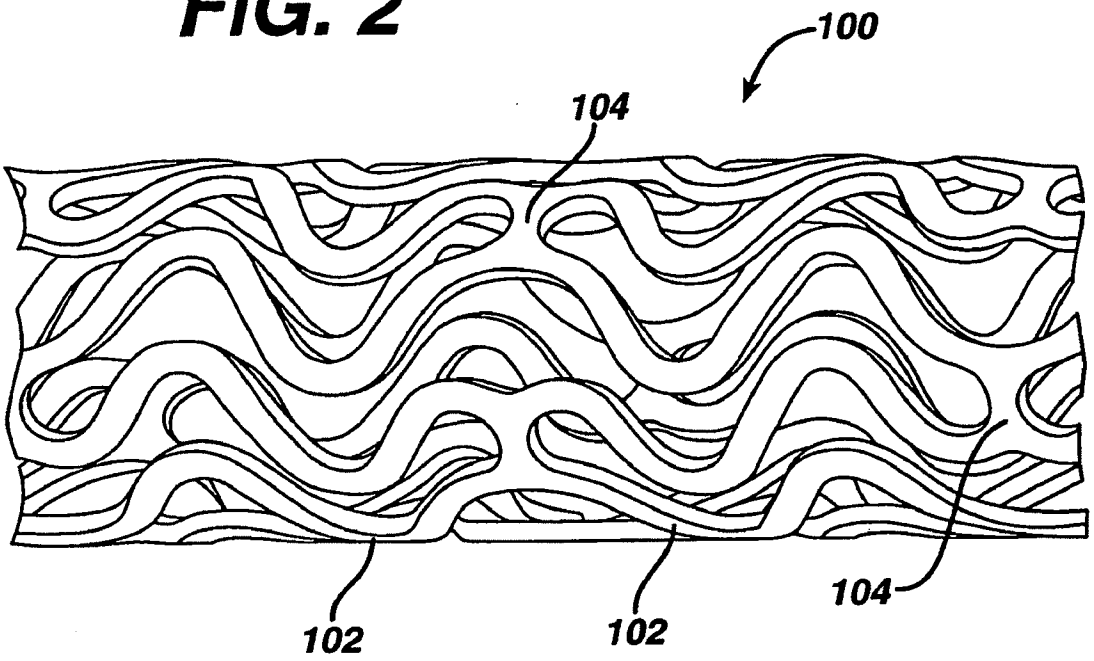
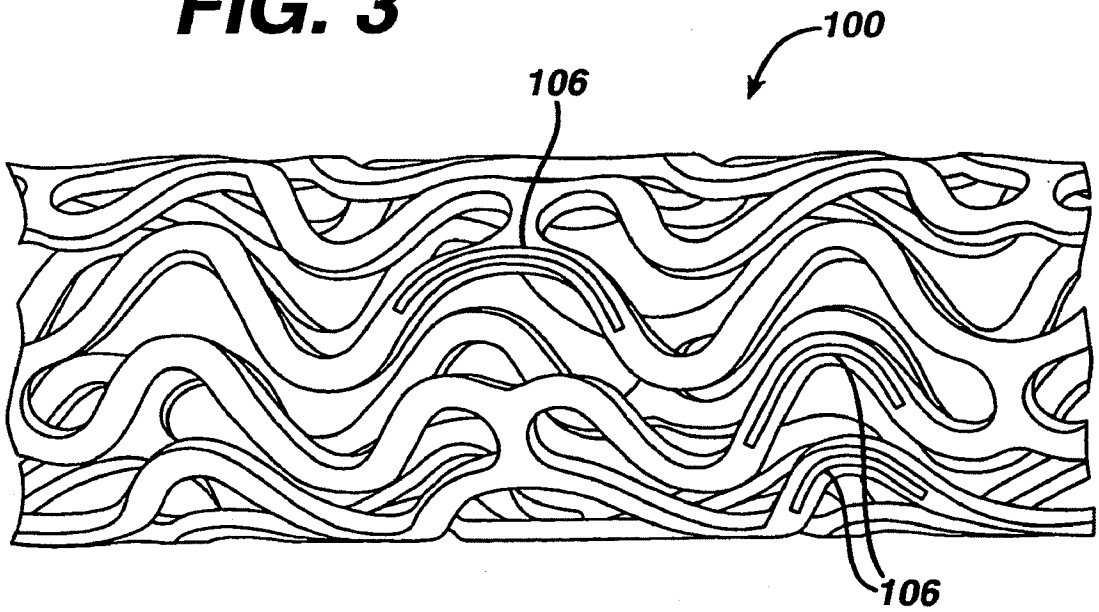


FIG. 3



DRUG/DRUG DELIVERY SYSTEMS FOR THE PREVENTION AND TREATMENT OF VASCULAR DISEASE

CROSS REFERENCE TO RELATED APPLICATIONS

This application is a continuation-in-part of U.S. application Ser. No. 09/850,293, filed May 7, 2001, now abandoned, which in turn claims priority of U.S. Provisional Application No. 60/263,979, filed Jan. 25, 2001, U.S. Provisional Application No. 60/263,806, filed January 24, 2001, U.S. Provisional Application No. 60/262,614, filed Jan. 18, 2001, U.S. Provisional Application No. 60/262,461, filed Jan. 18, 2001, and is a continuation-in-part of U.S. Application No. 09,575,480, filed May 19, 2000, now pending, which in turn claims priority of U.S. Provisional Application No. 60/204,4 17, filed May 12, 2000.

BACKGROUND OF THE INVENTION

1. Field of the Invention

The present invention relates to drugs and drug delivery systems for the prevention and treatment of vascular disease, and more particularly to drugs and drug delivery systems for the prevention and treatment of neointimal hyperplasia.

2. Discussion of the Related Art

Many individuals suffer from circulatory disease caused by a progressive blockage of the blood vessels that perfuse the heart and other major organs with nutrients. More severe blockage of blood vessels in such individuals often leads to hypertension, ischemic injury, stroke, or myocardial infarction. Atherosclerotic lesions, which limit or obstruct coronary blood flow, are the major cause of ischemic heart disease. Percutaneous transluminal coronary angioplasty is a medical procedure whose purpose is to increase blood flow through an artery. Percutaneous transluminal coronary angioplasty is the predominant treatment for coronary vessel stenosis. The increasing use of this procedure is attributable to its relatively high success rate and its minimal invasiveness compared with coronary bypass surgery. A limitation associated with percutaneous transluminal coronary angioplasty is the abrupt closure of the vessel which may occur immediately after the procedure and restenosis which occurs gradually following the procedure. Additionally, restenosis is a chronic problem in patients who have undergone saphenous vein bypass grafting. The mechanism of acute occlusion appears to involve several factors and may result from vascular recoil with resultant closure of the artery and/or deposition of blood platelets and fibrin along the damaged length of the newly opened blood vessel.

Restenosis after percutaneous transluminal coronary angioplasty is a more gradual process initiated by vascular injury. Multiple processes, including thrombosis, inflammation, growth factor and cytokine release, cell proliferation, cell migration and extracellular matrix synthesis each contribute to the restenotic process.

While the exact mechanism of restenosis is not completely understood, the general aspects of the restenosis process have been identified. In the normal arterial wall, smooth muscle cells proliferate at a low rate, approximately less than 0.1 percent per day. Smooth muscle cells in the vessel walls exist in a contractile phenotype characterized by eighty to ninety percent of the cell cytoplasmic volume occupied with the contractile apparatus. Endoplasmic reticulum, Golgi, and free ribosomes are few and are located in the perinuclear region. Extracellular matrix surrounds the

smooth muscle cells and is rich in heparin-like glycosylaminoglycans which are believed to be responsible for maintaining smooth muscle cells in the contractile phenotypic state (Campbell and Campbell, 1985).

Upon pressure expansion of an intracoronary balloon catheter during angioplasty, smooth muscle cells within the vessel wall become injured, initiating a thrombotic and inflammatory response. Cell derived growth factors such as platelet derived growth factor, fibroblast growth factor, epidermal growth factor, thrombin, etc., released from platelets, invading macrophages and/or leukocytes, or directly from the smooth muscle cells provoke proliferative and migratory responses in medial smooth muscle cells. These cells undergo a change from the contractile phenotype to a synthetic phenotype characterized by only a few contractile filament bundles, extensive rough endoplasmic reticulum, Golgi and free ribosomes. Proliferation/migration usually begins within one to two days post-injury and peaks several days thereafter (Campbell and Campbell, 1987; Clowes and Schwartz, 1985).

Daughter cells migrate to the intimal layer of arterial smooth muscle and continue to proliferate and secrete significant amounts of extracellular matrix proteins. Proliferation, migration and extracellular matrix synthesis continue until the damaged endothelial layer is repaired at which time proliferation slows within the intima, usually within seven to fourteen days post-injury. The newly formed tissue is called neointima. The further vascular narrowing that occurs over the next three to six months is due primarily to negative or constrictive remodeling.

Simultaneous with local proliferation and migration, inflammatory cells invade the site of vascular injury. Within three to seven days post-injury, inflammatory cells have migrated to the deeper layers of the vessel wall. In animal models employing either balloon injury or stent implantation, inflammatory cells may persist at the site of vascular injury for at least thirty days (Tanaka et al., 1993; Edelman et al., 1998). Inflammatory cells therefore are present and may contribute to both the acute and chronic phases of restenosis.

Numerous agents have been examined for presumed anti-proliferative actions in restenosis and have shown some activity in experimental animal models. Some of the agents which have been shown to successfully reduce the extent of intimal hyperplasia in animal models include: heparin and heparin fragments (Clowes, A. W. and Karnovsky M., Nature 265: 25-26, 1977; Guyton, J. R. et al., Circ. Res., 46: 625-634, 1980; Clowes, A. W. and Clowes, M. M., Lab. Invest. 52: 611-616, 1985; Clowes, A. W. and Clowes, M. M., Circ. Res. 58: 839-845, 1986; Majesky et al., Circ. Res. 61: 296-300, 1987; Snow et al., Am. J. Pathol. 137: 313-330, 1990; Okada, T. et al., Neurosurgery 25: 92-98, 1989), colchicine (Currier, J. W. et al., Circ. 80: 11-66, 1989), taxol (Sollot, S. J. et al., J. Clin. Invest. 95: 1869-1876, 1995), angiotensin converting enzyme (ACE) inhibitors (Powell, J. S. et al., Science, 245: 186-188, 1989), angiopeptin (Lundergan, C. F. et al. Am. J. Cardiol. 17(Suppl. B):132B-136B, 1991), cyclosporin A (Jonasson, L. et al., Proc. Natl., Acad. Sci., 85: 2303, 1988), goat-anti-rabbit PDGF antibody (Ferns, G. A. A., et al., Science 253: 1129-1132, 1991), terbinafine (Nemecek, G. M. et al., J. Pharmacol. Exp. Thera. 248: 1167-1174, 1989), trapidil (Liu, M. W. et al., Circ. 81: 1089-1093, 1990), tranilast (Fukuyama, J. et al., Eur. J. Pharmacol. 318: 327-332, 1996), interferon-gamma (Hansson, G. K. and Holm, J., Circ. 84: 1266-1272, 1991), rapamycin (Marx, S. O. et al., Circ. Res. 76: 412-417, 1995), corticosteroids (Colburn, M. D. et al., J. Vasc. Surg. 15:

510-518, 1992), see also Berk, B. C. et al., J. Am. Coll. Cardiol. 17: 111B-117B, 1991), ionizing radiation (Weinberger, J. et al., Int. J. Rad. Onc. Biol. Phys. 36: 767-775, 1996), fusion toxins (Farb, A. et al., Circ. Res. 80: 542-550, 1997) antisense oligonucleotides (Simons, M. et al., Nature 359: 67-70, 1992) and gene vectors (Chang, M. W. et al., J. Clin. Invest. 96: 2260-2268, 1995). Anti-proliferative effects on smooth muscle cells in vitro have been demonstrated for many of these agents, including heparin and heparin conjugates, taxol, tranilast, colchicine, ACE inhibitors, fusion toxins, antisense oligonucleotides, rapamycin and ionizing radiation. Thus, agents with diverse mechanisms of smooth muscle cell inhibition may have therapeutic utility in reducing intimal hyperplasia.

However, in contrast to animal models, attempts in human angioplasty patients to prevent restenosis by systemic pharmacologic means have thus far been unsuccessful. Neither aspirin-dipyridamole, ticlopidine, anti-coagulant therapy (acute heparin, chronic warfarin, hirudin or hirulog), thromboxane receptor antagonism nor steroids have been effective in preventing restenosis, although platelet inhibitors have been effective in preventing acute reocclusion after angioplasty (Mak and Topol, 1997; Lang et al., 1991; Popma et al., 1991). The platelet GP IIb/IIIa receptor, antagonist, Reopro is still under study but has not shown promising results for the reduction in restenosis following angioplasty and stenting. Other agents, which have also been unsuccessful in the prevention of restenosis, include the calcium channel antagonists, prostacyclin mimetics, angiotensin converting enzyme inhibitors, serotonin receptor antagonists, and anti-proliferative agents. These agents must be given systemically, however, and attainment of a therapeutically effective dose may not be possible; anti-proliferative (or anti-restenosis) concentrations may exceed the known toxic concentrations of these agents so that levels sufficient to produce smooth muscle inhibition may not be reached (Mak and Topol, 1997; Lang et al., 1991; Popma et al., 1991).

Additional clinical trials in which the effectiveness for preventing restenosis utilizing dietary fish oil supplements or cholesterol lowering agents has been examined showing either conflicting or negative results so that no pharmacological agents are as yet clinically available to prevent post-angioplasty restenosis (Mak and Topol, 1997; Franklin and Faxon, 1993: Serruys, P. W. et al., 1993). Recent observations suggest that the antilipid/antioxidant agent, probucol may be useful in preventing restenosis but this work requires confirmation (Tardif et al., 1997; Yokoi, et al., 1997). Probucol is presently not approved for use in the United States and a thirty-day pretreatment period would preclude its use in emergency angioplasty. Additionally, the application of ionizing radiation has shown significant promise in reducing or preventing restenosis after angioplasty in patients with stents (Teirstein et al., 1997). Currently, however, the most effective treatments for restenosis are repeat angioplasty, atherectomy or coronary artery bypass grafting, because no therapeutic agents currently have Food and Drug Administration approval for use for the prevention of post-angioplasty restenosis.

Unlike systemic pharmacologic therapy, stents have proven effective in significantly reducing restenosis. Typically, stents are balloon-expandable slotted metal tubes (usually, but not limited to, stainless steel), which, when expanded within the lumen of an angioplastied coronary artery, provide structural support through rigid scaffolding to the arterial wall. This support is helpful in maintaining vessel lumen patency. In two randomized clinical trials,

stents increased angiographic success after percutaneous transluminal coronary angioplasty, by increasing minimal lumen diameter and reducing, but not eliminating, the incidence of restenosis at six months (Serruys et al., 1994; Fischman et al., 1994).

Additionally, the heparin coating of stents appears to have the added benefit of producing a reduction in sub-acute thrombosis after stent implantation (Serruys et al., 1996). Thus, sustained mechanical expansion of a stenosed coronary artery with a stent has been shown to provide some measure of restenosis prevention, and the coating of stents with heparin has demonstrated both the feasibility and the clinical usefulness of delivering drugs locally, at the site of injured tissue.

Accordingly, there exists a need for effective drugs and drug delivery systems for the effective prevention and treatment of neointimal thickening that occurs after percutaneous transluminal coronary angioplasty and stent implantation.

SUMMARY OF THE INVENTION

The drugs and drug delivery systems of the present invention provide a means for overcoming the difficulties associated with the methods and devices currently in use as briefly described above.

In accordance with one aspect, the present invention is directed to a method for the prevention of constrictive remodeling. The method comprises the controlled delivery, by release from an intraluminal medical device, of a compound in therapeutic dosage amounts.

In accordance with another aspect, the present invention is directed to a drug delivery device. The drug delivery device comprises an intraluminal medical device and a therapeutic dosage of an agent releasably affixed to the intraluminal medical device for the treatment of constrictive vascular remodeling.

The drugs and drug delivery systems of the present invention utilize a stent or graft in combination with rapamycin or other drugs/agents/compounds to prevent and treat neointimal hyperplasia, i.e. restenosis, following percutaneous transluminal coronary angioplasty and stent implantation. It has been determined that rapamycin functions to inhibit smooth muscle cell proliferation through a number of mechanisms. It has also been determined that rapamycin eluting stent coatings produce superior effects in humans, when compared to animals, with respect to the magnitude and duration of the reduction in neointimal hyperplasia. Rapamycin administration from a local delivery platform also produces an anti-inflammatory effect in the vessel wall that is distinct from and complimentary to its smooth muscle cell anti-proliferative effect. In addition, it has also been demonstrated that rapamycin inhibits constrictive vascular remodeling in humans.

Other drugs, agents or compounds which mimic certain actions of rapamycin may also be utilized in combination with local delivery systems or platforms.

The local administration of drugs, agents or compounds to stented vessels have the additional therapeutic benefit of higher tissue concentration than that which would be achievable through the systemic administration of the same drugs, agents or compounds. Other benefits include reduced systemic toxicity, single treatment, and ease of administration. An additional benefit of a local delivery device and drug, agent or compound therapy may be to reduce the dose of the therapeutic drugs, agents or compounds and thus limit their toxicity, while still achieving a reduction in restenosis.

BRIEF DESCRIPTION OF THE DRAWINGS

The foregoing and other features and advantages of the invention will be apparent from the following, more particular description of preferred embodiments of the invention, as illustrated in the accompanying drawings.

FIG. **1** is a chart indicating the effectiveness of rapamycin as an anti-inflammatory relative to other anti-inflammatories.

FIG. **2** is a view along the length of a stent (ends not shown) prior to expansion showing the exterior surface of the stent and the characteristic banding pattern.

FIG. **3** is a perspective view of the stent of FIG. **1** having reservoirs in accordance with the present invention.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

As stated above, the proliferation of vascular smooth muscle cells in response to mitogenic stimuli that are released during balloon angioplasty and stent implantation is the primary cause of neointimal hyperplasia. Excessive neointimal hyperplasia can often lead to impairment of blood flow, cardiac ischemia and the need for a repeat intervention in selected patients in high risk treatment groups. Yet repeat revascularization incurs risk of patient morbidity and mortality while adding significantly to the cost of health care. Given the widespread use of stents in interventional practice, there is a clear need for safe and effective inhibitors of neointimal hyperplasia.

Rapamycin is a macroyclic triene antibiotic produced by streptomyces hygroscopicus as disclosed in U.S. Pat. No. 3,929,992. It has been found that rapamycin inhibits the proliferation of vascular smooth muscle cells in vivo. Accordingly, rapamycin may be utilized in treating intimal smooth muscle cell hyperplasia, restenosis and vascular occlusion in a mammal, particularly following either biologically or mechanically mediated vascular injury, or under conditions that would predispose a mammal to suffering such a vascular injury. Rapamycin functions to inhibit smooth muscle cell proliferation and does not interfere with the re-endothelialization of the vessel walls.

Rapamycin functions to inhibit smooth muscle cell proliferation through a number of mechanisms. In addition, rapamycin reduces the other effects caused by vascular injury, for example, inflammation. The operation and various functions of rapamycin are described in detail below. Rapamycin as used throughout this application shall include rapamycin, rapamycin analogs, derivatives and congeners that bind FKBP12 and possess the same pharmacologic properties as rapamycin.

Rapamycin reduces vascular hyperplasia by antagonizing smooth muscle proliferation in response to mitogenic signals that are released during angioplasty. Inhibition of growth factor and cytokine mediated smooth muscle proliferation at the late G1 phase of the cell cycle is believed to be the dominant mechanism of action of rapamycin. However, rapamycin is also known to prevent T-cell proliferation and differentiation when administered systemically. This is the basis for its immunosuppresive activity and its ability to prevent graft rejection.

The molecular events that are responsible for the actions of rapamycin, a known anti-proliferative, which acts to reduce the magnitude and duration of neointimal hyperplasia, are still being elucidated. It is known, however, that rapamycin enters cells and binds to a high-affinity cytosolic protein called FKBP12. The complex of rapamycin and

FKPB12 in turn binds to and inhibits a phosphoinositide (PI)-3 kinase called the "mammalian Target of Rapamycin" or TOR. TOR is a protein kinase that plays a key role in mediating the downstream signaling events associated with mitogenic growth factors and cytokines in smooth muscle cells and T lymphocytes. These events include phosphorylation of p27, phosphorylation of p70 s6 kinase and phosphorylation of 4BP-1, an important regulator of protein translation.

It is recognized that rapamycin reduces restenosis by inhibiting neointimal hyperplasia. However, there is evidence that rapamycin may also inhibit the other major component of restenosis, namely, negative remodeling. Remodeling is a process whose mechanism is not clearly understood but which results in shrinkage of the external elastic lamina and reduction in lumenal area over time, generally a period of approximately three to six months in humans.

Negative or constrictive vascular remodeling may be quantified angiographically as the percent diameter stenosis at the lesion site where there is no stent to obstruct the process. If late lumen loss is abolished in-lesion, it may be inferred that negative remodeling has been inhibited. Another method of determining the degree of remodeling involves measuring in-lesion external elastic lamina area using intravascular ultrasound (IVUS). Intravascular ultrasound is a technique that can image the external elastic lamina as well as the vascular lumen. Changes in the external elastic lamina proximal and distal to the stent from the post-procedural timepoint to four-month and twelve-month follow-ups are reflective of remodeling changes.

Evidence that rapamycin exerts an effect on remodeling comes from human implant studies with rapamycin coated stents showing a very low degree of restenosis in-lesion as well as in-stent. In-lesion parameters are usually measured approximately five millimeters on either side of the stent i.e. proximal and distal. Since the stent is not present to control remodeling in these zones which are still affected by balloon expansion, it may be inferred that rapamycin is preventing vascular remodeling.

The data in Table 1 below illustrate that in-lesion percent diameter stenosis remains low in the rapamycin treated groups, even at twelve months. Accordingly, these results support the hypothesis that rapamycin reduces remodeling.

TABLE 1.0

Angiographic In-Lesion Percent Diameter Stenosis (%, mean ± SD and "n=") In Patients Who Received a Rapamycin-Coated Stent

Coating Group	Post Placement	4–6 month Follow Up	12 month Follow Up
Brazil	10.6 ± 5.7 (30)	13.6 ± 8.6 (30)	22.3 ± 7.2 (15)
Netherlands	14.7 ± 8.8	22.4 ± 6.4	—

Additional evidence supporting a reduction in negative remodeling with rapamycin comes from intravascular ultrasound data that was obtained from a first-in-man clinical program as illustrated in Table 2 below.

TABLE 2.0

Matched IVUS data in Patients Who Received a Rapamycin-Coated Stent

IVUS Parameter	Post (n=)	4-Month Follow-Up (n=)	12-Month Follow-Up (n=)
Mean proximal vessel area (mm^2)	16.53 ± 3.53 (27)	16.31 ± 4.36 (28)	13.96 ± 2.26 (13)
Mean distal vessel area (mm^2)	13.12 ± 3.68 (26)	13.53 ± 4.17 (26)	12.49 ± 3.25 (14)

The data illustrated that there is minimal loss of vessel area proximally or distally which indicates that inhibition of negative remodeling has occurred in vessels treated with rapamycin-coated stents.

Other than the stent itself, there have been no effective solutions to the problem of vascular remodeling. Accordingly, rapamycin may represent a biological approach to controlling the vascular remodeling phenomenon.

It may be hypothesized that rapamycin acts to reduce negative remodeling in several ways. By specifically blocking the proliferation of fibroblasts in the vascular wall in response to injury, rapamycin may reduce the formation of vascular scar tissue. Rapamycin may also affect the translation of key proteins involved in collagen formation or metabolism.

Rapamycin used in this context includes rapamycin and all analogs, derivatives and congeners that bind FKBP12 and possess the same pharmacologic properties as rapamycin.

In a preferred embodiment, the rapamycin is delivered by a local delivery device to control negative remodeling of an arterial segment after balloon angioplasty as a means of reducing or preventing restenosis. While any delivery device may be utilized, it is preferred that the delivery device comprises a stent that includes a coating or sheath which elutes or releases rapamycin. The delivery system for such a device may comprise a local infusion catheter that delivers rapamycin at a rate controlled by the administrator.

Rapamycin may also be delivered systemically using an oral dosage form or a chronic injectible depot form or a patch to deliver rapamycin for a period ranging from about seven to forty-five days to achieve vascular tissue levels that are sufficient to inhibit negative remodeling. Such treatment is to be used to reduce or prevent restenosis when administered several days prior to elective angioplasty with or without a stent.

Data generated in porcine and rabbit models show that the release of rapamycin into the vascular wall from a nonerodible polymeric stent coating in a range of doses (35-430 ug/5-18 mm coronary stent) produces a peak fifty to fifty-five percent reduction in neointimal hyperplasia as set forth in Table 3 below. This reduction, which is maximal at about twenty-eight to thirty days, is typically not sustained in the range of ninety to one hundred eighty days in the porcine model as set forth in Table 4 below.

TABLE 3.0

Animal Studies with Rapamycin-coated stents. Values are mean ± Standard Error of Mean

Study	Duration	Stent[1]	Rapamycin	N	Neointimal Area (mm^2)	% Change From Polyme	% Change From Metal
Porcine							
98009	14 days	Metal		8	2.04 ± 0.17		
		1X + rapamycin	153 μg	8	1.66 ± 0.17*	-42%	-19%
		1X + TC300 + rapamycin	155 μg	8	1.51 ± 0.19*	-47%	-26%
99005	28 days	Metal		10	2.29 ± 0.21		
				9	3.91 ± 0.60**		
		1X + TC30 + rapamycin	130 μg	8	2.81 ± 0.34		+23%
		1X + TC100 + rapamycin	120 μg	9	2.62 ± 0.21		+14%
99006	28 days	Metal		12	4.57 ± 0.46		
		EVA/BMA 3X		12	5.02 ± 0.62		+10%
		1X + rapamycin	125 μg	11	2.84 ± 0.31* **	-43%	-38%
		3X + rapamycin	430 μg	12	3.06 ± 0.17* **	-39%	-33%
		3X + rapamycin	157 μg	12	2.77 ± 0.41* **	-45%	-39%
99011	28 days	Metal		11	3.09 ± 0.27		
				11	4.52 ± 0.37		
		1X + rapamycin	189 μg	14	3.05 ± 0.35		-1%
		3X + rapamycin/dex	182/363 μg	14	2.72 ± 0.71		-12%
99021	60 days	Metal		12	2.14 ± 0.25		
		1X + rapamycin	181 μg	12	2.95 ± 0.38		+38%
99034	28 days	Metal		8	5.24 ± 0.58		
		1X + rapamycin	186 μg	8	2.47 ± 0.33**		-53%
		3X + rapamycin/dex	185/369 μg	6	2.42 ± 0.64**		-54%
20001	28 days	Metal		6	1.81 ± 0.09		
		1X + rapamycin	172 μg	5	1.66 ± 0.44		-8%
20007							
	30 days	Metal		9	2.94 ± 0.43		
		1XTC + rapamycin	155 μg	10	1.40 ± 0.11*		-52%*
Rabbit							
99019	28 days	Metal		8	1.20 ± 0.07		

TABLE 3.0-continued

Animal Studies with Rapamycin-coated stents.
Values are mean ± Standard Error of Mean

Study	Duration	Stent[1]	Rapamycin	N	Neointimal Area (mm^2)	% Change From Polyme	% Change From Metal
		EVA/BMA 1X		10	1.26 ± 0.16		+5%
		1X + rapamycin	64 μg	9	0.92 ± 0.14	−27%	−23%
		1X + rapamycin	196 μg	10	0.66 ± 0.12* **	−48%	−45%
99020	28 days	Metal		12	1.18 ± 0.10		
		EVA/BMA 1X + rapamycin	197 μg	8	0.81 ± 0.16		−32%

[1]Stent nomenclature: EVA/BMA 1X, 2X, and 3X signifies approx. 500 μg, 1000 μg, and 1500 μg total mass (polymer + drug), respectively. TC, top coat of 30 μg, 100 μg, or 300 μg drug-free BMA; Biphasic; 2 × 1X layers of rapamycin in EVA/BMA spearated by a 100 μg drug-free BMA layer.
[2]0.25 mg/kg/d × 14 d preceeded by a loading dose of 0.5 mg/kg/d × 3 d prior to stent implantation.
*p < 0.05 from EVA/BMA control.
**p < 0.05 from Metal;
#Inflammation score: (0 = essentially no intimal involvement; 1 = <25% intima involved; 2 =≧25% intima involved; 3 = >50% intima involved).

TABLE 4.0

180 day Porcine Study with Rapamycin-coated stents.
Values are mean ± Standard Error of Mean

Study	Duration	Stent[1]	Rapamycin	N	Neointimal Area (mm^2)	% Change From Polyme	% Change From Metal	Inflammation Score #
20007	3 days	Metal		10	0.38 ± 0.06			1.05 ± 0.06
(ETP-2-002233-P)		1XTC + rapamycin	155 μg	10	0.29 ± 0.03		−24%	1.08 ± 0.04
	30 days	Metal		9	2.94 ± 0.43			0.11 ± 0.08
		1XTC + rapamycin	155 μg	10	1.40 ± 0.11*		−52%*	0.25 ± 0.10
	90 days	Metal		10	3.45 ± 0.34			0.20 ± 0.08
		1XTC + rapamycin	155 μg	10	3.03 ± 0.29		−12%	0.80 ± 0.23
		1X + rapamycin	171 μg	10	2.86 ± 0.35		−17%	0.60 ± 0.23
	180 days	Metal		10	3.65 ± 0.39			0.65 ± 0.21
		1XTC + rapamycin	155 μg	10	3.34 ± 0.31		−8%	1.50 ± 0.34
		1X + rapamycin	171 μg	10	3.87 ± 0.28		+6%	1.68 ± 0.37

The release of rapamycin into the vascular wall of a human from a nonerodible polymeric stent coating provides superior results with respect to the magnitude and duration of the reduction in neointimal hyperplasia within the stent as compared to the vascular walls of animals as set forth above.

Humans implanted with a rapamycin coated stent comprising rapamycin in the same dose range as studied in animal models using the same polymeric matrix, as described above, reveal a much more profound reduction in neointimal hyperplasia than observed in animal models, based on the magnitude and duration of reduction in neointima. The human clinical response to rapamycin reveals essentially total abolition of neointimal hyperplasia inside the stent using both angiographic and intravascular ultrasound measurements. These results are sustained for at least one year as set forth in Table 5 below.

TABLE 5.0

Patients Treated (N = 45 patients) with a Rapamycin-coated Stent

Effectiveness Measures	Sirolimus FIM (N = 45 Patients, 45 Lesions)	95% Confidence Limit
Procedure Success (QCA)	100.0% (45/45)	[92.1%, 100.0%]
4-month In-Stent Diameter Stenosis (%)		
Mean ± SD (N)	4.8% ± 6.1% (30)	[2.6%, 7.0%]
Range (min, max)	(−8.2%, 14.9%)	
6-month In-Stent Diameter Stenosis (%)		
Mean ± SD (N)	8.9% ± 7.6% (13)	[4.8%, 13.0%]
Range (min, max)	(−2.9%, 20.4%)	
12-month In-Stent Diameter Stenosis (%)		
Mean ± SD (N)	8.9% ± 6.1% (15)	[5.8%, 12.0%]
Range (min, max)	(−3.0%, 22.0%)	

TABLE 5.0-continued

Patients Treated (N = 45 patients) with a Rapamycin-coated Stent		
Effectiveness Measures	Sirolimus FIM (N = 45 Patients, 45 Lesions)	95% Confidence Limit
4-month In-Stent Late Loss (mm)		
Mean ± SD (N)	0.00 ± 0.29 (30)	[−0.10, 0.10]
Range (min, max)	(−0.51, 0.45)	
6-month In-Stent Late Loss (mm)		
Mean ± SD (N)	0.25 ± 0.27 (13)	[0.10, 0.39]
Range (min, max)	(−0.51, 0.91)	
12-month In-Stent Late Loss (mm)		
Mean ± SD (N)	0.11 ± 0.36 (15)	[−0.08, 0.29]
Range (min, max)	(−0.51, 0.82)	
4-month Obstruction Volume (%) (IVUS)		
Mean ± SD (N)	10.48% ± 2.78% (28)	[9.45%, 11.51%]
Range (min, max)	(4.60%, 16.35%)	
6-month Obstruction Volume (%) (IVUS)		
Mean ± SD (N)	7.22% ± 4.60% (13)	[4.72%, 9.72%],
Range (min, max)	(3.82%, 19.88%)	
12-month Obstruction Volume (%) (IVUS)		
Mean ± SD (N)	2.11% ± 5.28% (15)	[0.00%, 4.78%],
Range (min, max)	(0.00%, 19.89%)	
6-month Target Lesion Revascularization (TLR)	0.0% (0/30)	[0.0%, 9.5%]
12-month Target Lesion Revascularization (TLR)	0.0% (0/15)	[0.0%, 18.1%]

QCA = Quantitative Coronary Angiography
SD = Standard Deviation
IVUS = Intravascular Ultrasound

Rapamycin produces an unexpected benefit in humans when delivered from a stent by causing a profound reduction in in-stent neointimal hyperplasia that is sustained for at least one year. The magnitude and duration of this benefit in humans is not predicted from animal model data. Rapamycin used in this context includes rapamycin and all analogs, derivatives and congeners that bind FKBP12 and possess the same pharmacologic properties as rapamycin.

These results may be due to a number of factors. For example, the greater effectiveness of rapamycin in humans is due to greater sensitivity of its mechanism(s) of action toward the pathophysiology of human vascular lesions compared to the pathophysiology of animal models of angioplasty. In addition, the combination of the dose applied to the stent and the polymer coating that controls the release of the drug is important in the effectiveness of the drug.

As stated above, rapamycin reduces vascular hyperplasia by antagonizing smooth muscle proliferation in response to mitogenic signals that are released during angioplasty injury. Also, it is known that rapamycin prevents T-cell proliferation and differentiation when administered systemically. It has also been determined that rapamycin exerts a local inflammatory effect in the vessel wall when administered from a stent in low doses for a sustained period of time (approximately two to six weeks). The local anti-inflammatory benefit is profound and unexpected. In combination with the smooth muscle anti-proliferative effect, this dual mode of action of rapamycin may be responsible for its exceptional efficacy.

Accordingly, rapamycin delivered from a local device platform, reduces neointimal hyperplasia by a combination of anti-inflammatory and smooth muscle anti-proliferative effects. Rapamycin used in this context means rapamycin and all analogs, derivatives and congeners that bind FKBP12 and possess the same pharmacologic properties as rapamycin. Local device platforms include stent coatings, stent sheaths, grafts and local drug infusion catheters or porous balloons or any other suitable means for the in situ or local delivery of drugs, agents or compounds.

The anti-inflammatory effect of rapamycin is evident in data from an experiment, illustrated in Table 6, in which rapamycin delivered from a stent was compared with dexamethasone delivered from a stent. Dexamethasone, a potent steroidal anti-inflammatory agent, was used as a reference standard. Although dexamethasone is able to reduce inflammation scores, rapamycin is far more effective than dexamethasone in reducing inflammation scores. In addition, rapamycin significantly reduces neointimal hyperplasia, unlike dexamethasone.

TABLE 6.0

Group Rapamycin Rap	N=	Neointimal Area (mm^2)	% Area Stenosis	Inflammation Score
Uncoated	8	5.24 ± 1.65	54 ± 19	0.97 ± 1.00
Dexamethasone (Dex)	8	4.31 ± 3.02	45 ± 31	0.39 ± 0.24
Rapamycin (Rap)	7	2.47 ± 0.94*	26 ± 10*	0.13 ± 0.19*
Rap + Dex	6	2.42 ± 1.58*	26 ± 18*	0.17 ± 0.30*

*= significance level $P < 0.05$

Rapamycin has also been found to reduce cytokine levels in vascular tissue when delivered from a stent. The data in FIG. 1 illustrates that rapamycin is highly effective in reducing monocyte chemotactic protein (MCP-1) levels in

the vascular wall. MCP-1 is an example of a proinflammatory/chemotactic cytokine that is elaborated during vessel injury. Reduction in MCP-1 illustrates the beneficial effect of rapamycin in reducing the expression of proinflammatory mediators and contributing to the anti-inflammatory effect of rapamycin delivered locally from a stent. It is recognized that vascular inflammation in response to injury is a major contributor to the development of neointimal hyperplasia.

Since rapamycin may be shown to inhibit local inflammatory events in the vessel it is believed that this could explain the unexpected superiority of rapamycin in inhibiting neointima.

As set forth above, rapamycin functions on a number of levels to produce such desired effects as the prevention of T-cell proliferation, the inhibition of negative remodeling, the reduction of inflammation, and the prevention of smooth muscle cell proliferation. While the exact mechanisms of these functions are not completely known, the mechanisms that have been identified may be expanded upon.

Studies with rapamycin suggest that the prevention of smooth muscle cell proliferation by blockade of the cell cycle is a valid strategy for reducing neointimal hyperplasia. Dramatic and sustained reductions in late lumen loss and neointimal plaque volume have been observed in patients receiving rapamycin delivered locally from a stent. The present invention expands upon the mechanism of rapamycin to include additional approaches to inhibit the cell cycle and reduce neointimal hyperplasia without producing toxicity.

The cell cycle is a tightly controlled biochemical cascade of events that regulate the process of cell replication. When cells are stimulated by appropriate growth factors, they move from G_0 (quiescence) to the G1 phase of the cell cycle. Selective inhibition of the cell cycle in the G1 phase, prior to DNA replication (S phase), may offer therapeutic advantages of cell preservation and viability while retaining anti-proliferative efficacy when compared to therapeutics that act later in the cell cycle i.e. at S, G2 or M phase.

Accordingly, the prevention of intimal hyperplasia in blood vessels and other conduit vessels in the body may be achieved using cell cycle inhibitors that act selectively at the G1 phase of the cell cycle. These inhibitors of the G1 phase of the cell cycle may be small molecules, peptides, proteins, oligonucleotides or DNA sequences. More specifically, these drugs or agents include inhibitors of cyclin dependent kinases (cdk's) involved with the progression of the cell cycle through the G1 phase, in particular cdk2 and cdk4.

Examples of drugs, agents or compounds that act selectively at the G1 phase of the cell cycle include small molecules such as flavopiridol and its structural analogs that have been found to inhibit cell cycle in the late G1 phase by antagonism of cyclin dependent kinases. Therapeutic agents that elevate an endogenous kinase inhibitory protein^{kip} called P27, sometimes referred to as P27^{kip1}, that selectively inhibits cyclin dependent kinases may be utilized. This includes small molecules, peptides and proteins that either block the degradation of P27 or enhance the cellular production of P27, including gene vectors that can transfact the gene to produce P27. Staurosporin and related small molecules that block the cell cycle by inhibiting protein kinases may be utilized. Protein kinase inhibitors, including the class of tyrphostins that selectively inhibit protein kinases to antagonize signal transduction in smooth muscle in response to a broad range of growth factors such as PDGF and FGF may also be utilized.

Any of the drugs, agents or compounds discussed above may be administered either systemically, for example, orally, intravenously, intramuscularly, subcutaneously, nasally or intradermally, or locally, for example, stent coating, stent covering or local delivery catheter. In addition, the drugs or agents discussed above may be formulated for fast-release or slow release with the objective of maintaining the drugs or agents in contact with target tissues for a period ranging from three days to eight weeks.

As set forth above, the complex of rapamycin and FKPB12 binds to and inhibits a phosphoinositide (PI)-3 kinase called the mammalian Target of Rapamycin or TOR. An antagonist of the catalytic activity of TOR, functioning as either an active site inhibitor or as an allosteric modulator, i.e. an indirect inhibitor that allosterically modulates, would mimic the actions of rapamycin but bypass the requirement for FKBP12. The potential advantages of a direct inhibitor of TOR include better tissue penetration and better physical/chemical stability. In addition, other potential advantages include greater selectivity and specificity of action due to the specificity of an antagonist for one of multiple isoforms of TOR that may exist in different tissues, and a potentially different spectrum of downstream effects leading to greater drug efficacy and/or safety.

The inhibitor may be a small organic molecule (approximate mw<1000), which is either a synthetic or naturally derived product. Wortmanin may be an agent which inhibits the function of this class of proteins. It may also be a peptide or an oligonucleotide sequence. The inhibitor may be administered either sytemically (orally, intravenously, intramuscularly, subcutaneously, nasally, or intradermally) or locally (stent coating, stent covering, local drug delivery catheter). For example, the inhibitor may be released into the vascular wall of a human from a nonerodible polymeric stent coating. In addition, the inhibitor may be formulated for fast-release or slow release with the objective of maintaining the rapamycin or other drug, agent or compound in contact with target tissues for a period ranging from three days to eight weeks.

As stated previously, the implantation of a coronary stent in conjunction with balloon angioplasty is highly effective in treating acute vessel closure and may reduce the risk of restenosis. Intravascular ultrasound studies (Mintz et al., 1996) suggest that coronary stenting effectively prevents vessel constriction and that most of the late luminal loss after stent implantation is due to plaque growth, probably related to neointimal hyperplasia. The late luminal loss after coronary stenting is almost two times higher than that observed after conventional balloon angioplasty. Thus, inasmuch as stents prevent at least a portion of the restenosis process, the use of drugs, agents or compounds which prevent inflammation and proliferation, or prevent proliferation by multiple mechanisms, combined with a stent may provide the most efficacious treatment for post-angioplasty restenosis.

The local delivery of drugs, agents or compounds from a stent has the following advantages; namely, the prevention of vessel recoil and remodeling through the scaffolding action of the stent and the drugs, agents or compounds and the prevention of multiple components of neointimal hyperplasia. This local administration of drugs, agents or compounds to stented coronary arteries may also have additional therapeutic benefit. For example, higher tissue concentrations would be achievable than that which would occur with systemic administration, reduced systemic toxicity, and single treatment and ease of administration. An additional benefit of drug therapy may be to reduce the dose of the therapeutic compounds, thereby limiting their toxicity, while still achieving a reduction in restenosis.

There are a multiplicity of different stents that may be utilized following percutaneous transluminal coronary angioplasty. Although any number of stents may be utilized in accordance with the present invention, for simplicity, one particular stent will be described in exemplary embodiments of the present invention. The skilled artisan will recognize that any number of stents may be utilized in connection with the present invention.

A stent is commonly used as a tubular structure left inside the lumen of a duct to relieve an obstruction. Commonly, stents are inserted into the lumen in a non-expanded form and are then expanded autonomously, or with the aid of a second device in situ. A typical method of expansion occurs through the use of a catheter-mounted angioplasty balloon which is inflated within the stenosed vessel or body passageway in order to shear and disrupt the obstructions associated with the wall components of the vessel and to obtain an enlarged lumen. As set forth below, self-expanding stents may also be utilized.

FIG. **2** illustrates an exemplary stent **100** which may be utilized in accordance with an exemplary embodiment of the present invention. The expandable cylindrical stent **100** comprises a fenestrated structure for placement in a blood vessel, duct or lumen to hold the vessel, duct or lumen open, more particularly for protecting a segment of artery from restenosis after angioplasty. The stent **100** may be expanded circumferentially and maintained in an expanded configuration, that is circumferentially or radially rigid. The stent **100** is axially flexible and when flexed at a band, the stent **100** avoids any externally-protruding component parts.

The stent **100** generally comprises first and second ends with an intermediate section therebetween. The stent **100** has a longitudinal axis and comprises a plurality of longitudinally disposed bands **102**, wherein each band **102** defines a generally continuous wave along a line segment parallel to the longitudinal axis. A plurality of circumferentially arranged links **104** maintain the bands **102** in a substantially tubular structure. Essentially, each longitudinally disposed band **102** is connected at a plurality of periodic locations, by a short circumferentially arranged link **104** to an adjacent band **102**. The wave associated with each of the bands **102** has approximately the same fundamental spatial frequency in the intermediate section, and the bands **102** are so disposed that the wave associated with them are generally aligned so as to be generally in phase with one another. As illustrated in the figure, each longitudinally arranged band **102** undulates through approximately two cycles before there is a link to an adjacent band.

The stent **100** may be fabricated utilizing any number of methods. For example, the stent **100** may be fabricated from a hollow or formed stainless steel tube that may be machined using lasers, electric discharge milling, chemical etching or other means. The stent **100** is inserted into the body and placed at the desired site in an unexpanded form. In one embodiment, expansion may be effected in a blood vessel by a balloon catheter, where the final diameter of the stent **100** is a function of the diameter of the balloon catheter used.

It should be appreciated that a stent **100** in accordance with the present invention may be embodied in a shape-memory material, including, for example, an appropriate alloy of nickel and titanium. In this embodiment, after the stent **100** has been formed it may be compressed so as to occupy a space sufficiently small as to permit its insertion in a blood vessel or other tissue by insertion means, wherein the insertion means include a suitable catheter, or flexible rod. On emerging from the catheter, the stent **100** may be configured to expand into the desired configuration where the expansion is automatic or triggered by a change in pressure, temperature or electrical stimulation.

FIG. **3** illustrates an exemplary embodiment of the present invention utilizing the stent **100** illustrated in FIG. **2**. As illustrated, the stent **100** may be modified to comprise a reservoir **106**. Each of the reservoirs may be opened or closed as desired. These reservoirs **106** may be specifically designed to hold the drug, agent, compound or combinations thereof to be delivered. Regardless of the design of the stent **100**, it is preferable to have the drug, agent, compound or combinations thereof dosage applied with enough specificity and a sufficient concentration to provide an effective dosage in the lesion area. In this regard, the reservoir size in the bands **102** is preferably sized to adequately apply the drug/drug combination dosage at the desired location and in the desired amount.

In an alternate exemplary embodiment, the entire inner and outer surface of the stent **100** may be coated with various drug and drug combinations in therapeutic dosage amounts. A detailed description of exemplary coating techniques is described below.

Rapamycin or any of the drugs, agents or compounds described above may be incorporated into or affixed to the stent in a number of ways and utilizing any number of biocompatible materials. In the exemplary embodiment, the rapamycin is directly incorporated into a polymeric matrix and sprayed onto the outer surface of the stent. The rapamycin elutes from the polymeric matrix over time and enters the surrounding tissue. The rapamycin preferably remains on the stent for at least three days up to approximately six months and more preferably between seven and thirty days.

Any number of non-erodible polymers may be utilized in conjunction with rapamycin. In the exemplary embodiment, the polymeric matrix comprises two layers. The base layer comprises a solution of ethylene-co-vinylacetate and polybutylmethacrylate. The rapamycin is incorporated into this layer. The outer layer comprises only polybutylmethacrylate and acts as a diffusion barrier to prevent the rapamycin from eluting too quickly and entering the surrounding tissues. The thickness of the outer layer or top coat determines the rate at which the rapamycin elutes from the matrix. Essentially, the rapamycin elutes from the matrix by diffusion through the polymer molecules. Polymers tend to move, thereby allowing solids, liquids and gases to escape therefrom. The total thickness of the polymeric matrix is in the range from about 1 micron to about 20 microns or greater. In a preferred exemplary embodiment, the base layer, including the polymer and drug, has a thickness in the range from about 8 microns to about 12 microns and the outer layer has a thickness in the range from about 1 micron to about 2 microns.

The ethylene-co-vinylacetate, polybutylmethacrylate and rapamycin solution may be incorporated into or onto the stent in a number of ways. For example, the solution may be sprayed onto the stent or the stent may be dipped into the solution. In a preferred embodiment, the solution is sprayed onto the stent and then allowed to dry. In another exemplary embodiment, the solution may be electrically charged to one polarity and the stent electrically changed to the opposite polarity. In this manner, the solution and stent will be attracted to one another. In using this type of spraying process, waste may be reduced and more control over the thickness of the coat may be achieved.

Since rapamycin works by entering the surrounding tissue, it is preferably only affixed to the surface of the stent making contact with one tissue. Typically, only the outer surface of the stent makes contact with the tissue. Accord-

ingly, in a preferred embodiment, only the outer surface of the stent is coated with rapamycin. For other drugs, agents or compounds, the entire stent may be coated.

It is important to note that different polymers may be utilized for different stents. For example, in the above-described embodiment, ethylene-co-vinylacetate and polybutylmethacrylate are utilized to form the polymeric matrix. This matrix works well with stainless steel stents. Other polymers may be utilized more effectively with stents formed from other materials, including materials that exhibit superelastic properties such as alloys of nickel and titanium.

Although shown and described is what is believed to be the most practical and preferred embodiments, it is apparent that departures from specific designs and methods described and shown will suggest themselves to those skilled in the art and may be used without departing from the spirit and scope of the invention. The present invention is not restricted to the particular constructions described and illustrated, but should be constructed to cohere with all modifications that may fall within the scope of the appended claims.

What is claimed is:

1. A drug delivery device comprising: an intraluminal stent; a biocompatible, nonerodible polymeric coating affixed to the intraluminal stent; and from about 64 μg to about 197 μg of rapamycin or a macrocyclic triene analog thereof that binds FKBP12 incorporated into the polymeric coating, wherein said device provides an in-stent late loss in diameter at 12 months following implantation in a human of less than about 0.5 mm, as measured by quantitative coronary angiography.

2. A drug delivery device according to claim **1** that provides an in- stent late loss in diameter at 12 months following implantation in a human of less than about 0.3 mm, as measured by quantitative coronary angiography.

3. A drug delivery device according to claim **1** or **2** that provides an in-stent diameter stenosis at 12 months following implantation in a human of less than about 22%, as measured by quantitative coronary angiography.

4. A drug delivery device according to claim **3** that provides an in-stent diameter stenosis at 12 months following implantation in a human of less than about 15%, as measured by quantitative coronary angiography.

5. A drug delivery device comprising: an intraluminal stent; a biocompatible, nonerodible polymeric coating affixed to the intraluminal stent; and from about 64 μg to about 197 μg of rapamycin or a macrocyclic triene analog thereof that binds FKBP12 incorporated into the polymeric coating, wherein said device provides a mean in-stent late loss in diameter in a human population at 12 months following implantation of less than about 0.5 mm, as measured by quantitative coronary angiography.

6. A drug delivery device according to claim **5** that provides a mean in-stent late loss in diameter in a human population at 12 months following implantation of less than about 0.3 mm, as measured by quantitative coronary angiography.

7. A drug delivery device according to claim **5** or **6** that provides a mean in-stent diameter stenosis in a human population at 12 months following implantation of less than about 22%, as measured by quantitative coronary angiography.

8. A drug delivery device according to claim **7** that provides a mean in-stent diameter stenosis in a human population at 12 months following implantation of less than about 15%, as measured by quantitative coronary angiography.

9. A method of inhibiting neointimal proliferation in a human coronary artery resulting from percutaneous transluminal coronary angioplasty comprising implanting in the lumen of said coronary artery a drug delivery device comprising: an intraluminal stent; a biocompatible, nonerodible polymeric coating affixed to the intraluminal stent; and from about 64 μg to about 197 μg of rapamycin or a macrocyclic triene analog thereof that binds FKBP12 incorporated into the polymeric coating, wherein said method provides an in-stent late loss in diameter at 12 months following implantation of less than about 0.5 mm, as measured by quantitative coronary angiography.

10. A method according to claim **9** that provides an in-stent late loss in diameter at 12 months following implantation of less than about 0.3 mm, as measured by quantitative coronary angiography.

11. A method according to claim **9** or **10** that provides an in-stent diameter stenosis at 12 months following implantation of less than about 22%, as measured by quantitative coronary angiography.

12. A method according to claim **11** that provides an in-stent diameter stenosis at 12 months following implantation of less than about 15%, as measured by quantitative coronary angiography.

13. A method of inhibiting neointimal proliferation in a coronary artery resulting from percutaneous transluminal coronary angioplasty comprising implanting in the lumen of said coronary artery a drug delivery device comprising: an intraluminal stent; a biocompatible, nonerodible polymeric coating affixed to the intraluminal stent; and from about 64 μg to about 197 μg of rapamycin or a macrocyclic triene analog thereof that binds FKBP12 incorporated into the polymeric coating, wherein said method provides a mean in-stent late loss in diameter in a human population at 12 months following implantation of less than about 0.5 mm, as measured by quantitative coronary angiography.

14. A method according to claim **13** that provides a mean in-stent late loss in diameter in a human population at 12 months following implantation of less than about 0.3 mm, as measured by quantitative coronary angiography.

15. A method according to claim **13** or **14** that provides a mean in-stent diameter stenosis in a human population at 12 months following implantation of less than about 22%, as measured by quantitative coronary angiography.

16. A method according to claim **15** that provides a mean in-stent diameter stenosis in a human population at 12 months following implantation of less than about 15%, as measured by quantitative coronary angiography.

17. The drug delivery device according to any one of claims **1**, **2**, **4** or **5** wherein said rapamycin or macrocyclic triene analog thereof is incorporated into the polymeric coating at a dose of from about 64 μg to about 125 μg.

18. The drug delivery device according to to any one of claims **1**, **2**, **4** or **5** that releases a portion of said dose of rapamycin or a macrocyclic triene analog thereof at about six weeks following intraluminal implantation.

19. The drug delivery device according to any one of claims **1**, **2**, **4** or **5** wherein said rapamycin or macrocyclic triene analog thereof is incorporated into the polymeric coating at a dose of from about 2 μg to about 30 μg per millimeter of stent length.

20. The drug delivery device according to claim **19** wherein said rapamycin or macrocyclic triene analog thereof is incorporated into the polymeric coating at a dose of from about 3 μg to about 13 μg per millimeter of stent length.

21. The drug delivery device according to claim **19** that releases a portion of said dose of rapamycin or a macrocyclic triene analog thereof at about six weeks following intraluminal implantation.

22. The method according to any one of claims **9**, **10**, **13** or **14**, wherein said rapamycin or macrocyclic triene analog thereof is incorporated into the polymeric coating at a dose of from about 64 μg to about 125 μg.

23. The method according to any one of claims **9**, **10**, **13** or **14**, wherein said rapamycin or macrocyclic triene analog thereof is incorporated into the polymeric coating at a dose of from about 2 μg to about 30 μg per millimeter of stent length.

24. The method according to any one of claims **9**, **10**, **13** or **14**, wherein said rapamycin or macrocyclic triene analog thereof is incorporated into the polymeric coating at a dose of from about 3 μg to about 13 μg per millimeter of stent length.

25. The method according to any one of claims **9**, **10**, **13** or **14**, wherein said drug delivery device releases a portion of said dose of rapamycin or a macrocyclic triene analog thereof at about six weeks following intraluminal implantation.

* * * * *

ELSEVIER

The In Vivo Effect of Rapamycin Derivative SDZ RAD on Lymphocyte Proliferation

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THE rapamycin derivative SDZ RAD is a new, orally applicable immunosuppressive agent. The 2-hydroxyethyl chain in position 40 increases the polarity of this rapamycin analogue, resulting in a greater oral bioavailability.[1] SDZ RAD efficiently prevents graft rejection in animal models of allotransplantation. In addition, SDZ RAD inhibits the development of long-term vascular changes associated with chronic renal allograft rejection. Like cyclosporine (CyA) and FK 506, both SDZ RAD and rapamycin exert their immunosuppressive actions by binding to a class of cytosolic proteins called *immunophyllines*. The abundant immunophyllin FKBP-12 is the common target of SDZ RAD, rapamycin, and FK 506. However, concerning the mode of action, SDZ RAD differs from CyA and FK 506. Whereas CyA and FK 506 prevent progression from G_0 to G_1 phase in antigen-stimulated T cells by inhibition of IL-2 synthesis, SDZ RAD inhibits cell-cycle progression from G_1 to S phase by blocking IL-2 receptor and CD28-dependent signalling pathways.[2-4] Recently it was shown in vitro and in different animal models that SDZ RAD acts synergistically together with CyA.[5] It was the purpose of the present study to investigate whether SDZ RAD has additional effects on T-cell proliferation in human renal allograft recipients treated with CyA and methylprednisolone.

MATERIALS AND METHODS

The in vivo effect of SDZ RAD was investigated in nine renal allograft recipients with stable graft function. Each patient received a single dose of SDZ RAD (0.75 mg: $n = 3$; 2.5 mg: $n = 3$; 7.5 mg: $n = 2$; 17.5 mg: $n = 1$). Additionally, four patients were treated with placebo. Maintenance immunosuppression consisted of CyA (trough level 100 to 150 ng/mL) and methylprednisolone (<12 mg/d). Co-medication did not change throughout the study. Additionally, 10 mL citrate blood was taken before, 2 hours, 6 hours, and 10 hours after the intake of medication. The day before (without SDZ RAD/placebo), control samples were taken at identical time points. Peripheral blood mononuclear cells (PBMCs) were isolated by Ficoll density centrifugation and seeded in 96-well plates at a final concentration of 1×10^5 cells/well. RPMI-1640 medium was supplemented with 10% FCS, 100 U penicillin/streptomycin (100 μg/mL), 2 mmol/L glutamine (Biochrom, Germany) and T-lymphocytes were stimulated with monoclonal anti-CD3 antibodies (Sigma, 10 ng/mL)[6] and incubated in humidified air at 37°C, 5% CO_2, for 72 hours. In addition, 20 μL solution (Sigma, 5 mg/mL) was added for another 5 hours. Then, 100 μL solution (10% in 0.01 mol/L HCl) was added to each well. After overnight incubation, the absorbance at 570 nm (A_{570}) was determined by ELISA reader.[7]

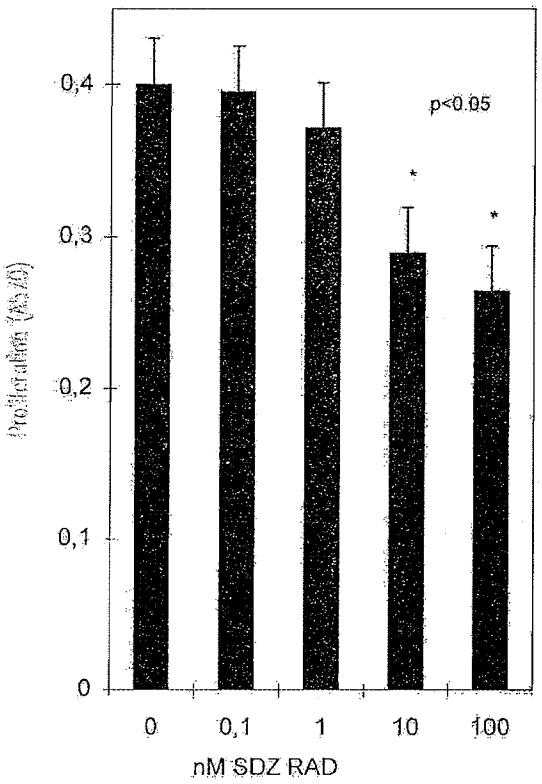

Fig 1. Effect of SDZ RAD on T-lymphocyte proliferation (mean ± SD) in vitro.

For in vitro experiments blood was taken from three different

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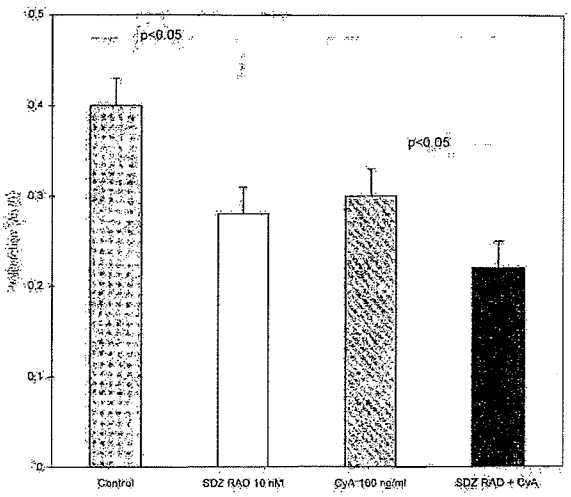

Fig 2. Treatment of PBMCs with a combination of SDZ RAD and CyA in vitro.

healthy volunteers. Stock solutions of SDZ RAD and CyA were prepared with EtOH and diluted to a final concentration in RPMI medium, as indicated in each single experiment. Lymphocyte proliferation was measured as outlined above, with different concentrations of SDZ RAD and CyA. Every experiment was done in triplicate. Intratest standard deviation was less then 1%. Statistics were evaluated by paired *t*-test. Data are expressed as mean ± standard error (SE). A probability of less than .05 was considered as statistically significant.

RESULTS

In vitro, the immunosuppressive activity of SDZ RAD on T-cell proliferation was assessed in cultures of anti-CD3-stimulated PBMCs derived from three healthy volunteers. SDZ RAD caused a significant and dose-dependent reduction of T-cell proliferation (Fig 1). SDZ RAD (10 nmol/L) suppressed proliferation from A_{570} = 0.4 ± 0.03 to 0.27 ± 0.03. Treatment of PBMCs with a combination of SDZ RAD and CyA revealed an additive effect of both agents on anti-CD3-driven T-cell proliferation (Fig 2).

In vivo, T-cell proliferation was significantly reduced by SDZ RAD (Fig 3). After intake, T-cell proliferation decreased at 2 hours (A_{570} = 0.41 vs 0.5, P = .01) and at 6 hours (A_{570} = 0.4 vs 0.5, P = .01) and at 10 hours the effect disappeared and proliferation returned to normal (A_{570} = 0.5 vs 0.5, NS). No such changes were found in patients who received placebo (data not shown).

The inhibition of T-cell proliferation seemed to be dose dependent, although this was not statistically significant owing to the low number of patients in each group (0.75 mg: ΔA_{570} = 0.07; 2.5 mg: ΔA_{570} = 0.15; 7.5 mg: ΔA_{570} = 0.12; 17.5 mg: ΔA_{570} = 0.27).

DISCUSSION

The immunosuppressive effects of CyA and methylprednisolone on lymphocyte proliferation can be enhanced by a

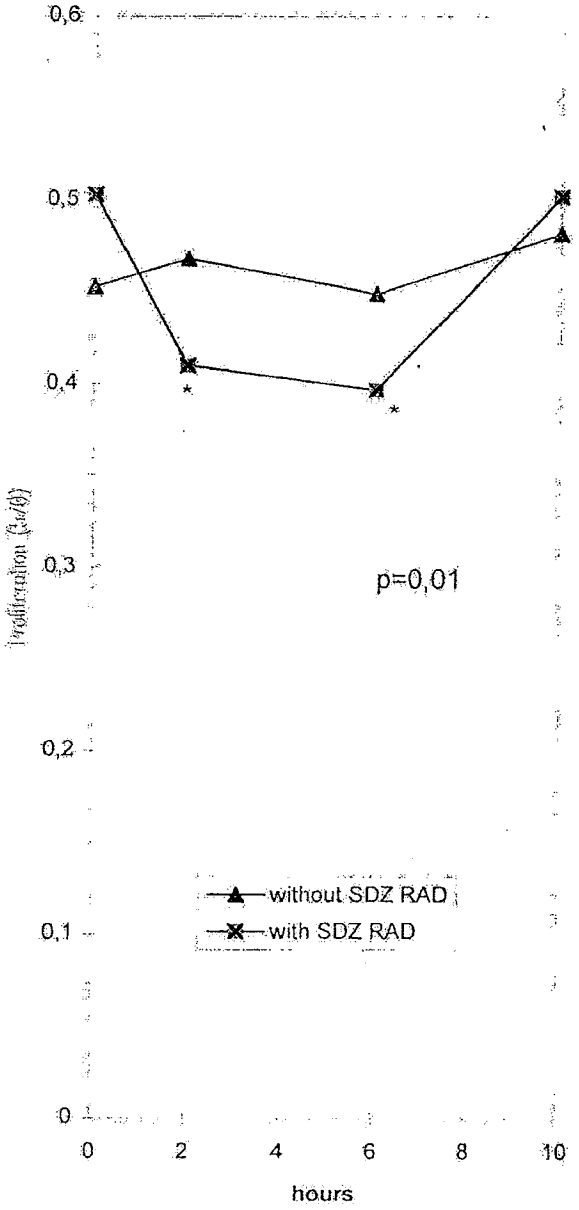

Fig 3. Effect of SDZ RAD on T-lymphocyte proliferation in vivo.

single dose of SDZ RAD in human renal allograft recipients. The inhibitory effect of SDZ RAD on the proliferation of T-lymphocytes reaches a maximum between 2 and 6 hours after intake and disappears at 10 hours after intake. Our results support the idea that SDZ RAD acts synergistically together with CyA in vitro and in vivo. Therefore, the combination of SDZ RAD with CyA may be beneficial. Further studies investigating the long-term effects of SDZ RAD on lymphocyte proliferation are under way.

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XIENCE V™ Stent Design and Rationale

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Drug-eluting stents (DES) are a preferred treatment modality for occlusive coronary artery disease. First-generation DES have demonstrated high levels of efficacy. However, concerns have been raised over late thrombotic events. XIENCE V™ everolimus-eluting coronary stent is a second-generation DES designed to be more deliverable and safe, while maintaining efficacy in a broad patient population compared with first-generation DES.[1–3] As a drug/device combination product, the overall performance of a DES is determined by its components and how well they are integrated. XIENCE V utilizes the MULTI-LINK VISION® stent, the antiproliferative drug everolimus, a fluorinated polymer drug carrier, poly(vinylidene fluoride-co-hexafluoropropylene) (PVDF-HFP), and a stent-specific delivery system. A DES coating must fulfill the multiple goals of biocompatibility, controlled drug release and maintenance of the coating durability through stent crimping, and expansion in vivo. The XIENCE V coating utilizes a two-layer coating system composed of an acrylate primer and a fluorinated copolymer drug reservoir. Fluorinated polymers have a long history of use in permanent vascular implant applications. The XIENCE V fluorinated copolymer offers in vivo biocompatibility combined with excellent chemical stability and high purity. Described in this article are the design rationale and polymer selection criteria. The hemocompatibility and biocompatibility of the fluorinated polymer coating are discussed. Characterization results on drug release control, possible drug release mechanism, coating integrity, coating uniformity, and fatigue resistance are also presented. (J Interven Cardiol 2009;22:S18–S27)

Introduction

The rapid acceptance of first-generation drug-eluting stents (DES) when introduced, combined with the more recent concerns over late stent thrombosis, has driven the industry to develop second-generation DES. The goals for second-generation DES are to offer an optimal balance of safety, effectiveness, and deliverability in broad patient populations. Abbott Vascular's XIENCE V everolimus-eluting coronary stent system was designed with the intent of meeting these goals. Through extensive polymer screening, multiple coating design iterations, and extensive preclinical studies, the XIENCE V DES configuration was finalized with four major components, that is, the proven MULTI-LINK VISION stent and stent delivery system, the antiproliferative agent everolimus, and a fluorinated copolymer drug carrier.

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Stent and Stent Delivery System

The MULTI-LINK VISION stent is made from a cobalt-chromium (Co-Cr) alloy. With a strut thickness of 81 μm, it has thinner struts than a stainless steel stent while still achieving excellent radiopacity and radial strength. The open cell and nonlinear link design makes the stent flexible and conformable to the vessel. Due to the thin struts, and optimized delivery system, the MULTI-LINK VISION system has a low profile at 0.040″ (1.0 mm) for a 3.0 × 18 mm stent, which enhances deliverability. The ML VISION stent delivery balloon is made of semicompliant polyether block amide (PEBAX®) material with short tapers and is designed to minimize injury outside of the stented region.

Drug

The drug everolimus, manufactured by Novartis Pharma AG, is a potent antiproliferative agent that acts on a wide range of cell types, including vascular smooth muscle cells, which it inhibits at a low nanomolar level (IC_{50} = 0.9–3.6 nM for bovine smooth muscle cells) in the G_1 phase of the cell cycle.[4] Everolimus is lipophilic, with an octanol/water partition coefficient of approximately 10.[4] The combination of smooth muscle cell inhibition, high potency, and high lipophilicity makes the drug an attractive candidate for controlled delivery from a DES for the prevention of neointimal hyperplasia. Multiple preclinical studies led to the selection of a low drug dose of 100 $\mu g/cm^2$ for the XIENCE V system. The XIENCE V stent releases approximately 80% of the drug in 1 month with near 100% drug release by 4 months. Details of preclinical pharmacokinetics are discussed in a separate article in this supplement issue. This drug dose is the lowest of all the commercialized "olimus" DES products.

XIENCE V Coating

The coating polymer plays a pivotal role in DES. Many factors need to be considered in selecting a polymer or polymer system. These include the hemocompatibility and biocompatibility of the polymer, polymer purity, coating integrity, polymer-drug miscibility, polymer processability, ability to control the drug release, and the stability (shelf-life) of the system. The XIENCE V system utilizes a two-layer coating construct composed of a primer layer and a drug-polymer reservoir layer with no topcoat. Poly(n-butyl methacrylate) (PBMA) is used as the thin, primer adhesion layer, and the drug reservoir layer is composed of poly(vinylidene fluoride-co-hexafluoropropylene) (PVDF-HFP) combined with everolimus. This two-layer coating construct is designed to ensure excellent adhesion of the drug matrix to the stent, while minimizing unwanted adhesions to the delivery balloon. The drug reservoir layer holds the drug onto the stent, controls the drug release, and contributes to the blood and vascular tissue compatibility of the DES stent. The following discussion will focus on the design rationale, the biocompatibility of the system, and the bench characterization results.

XIENCE V Formulation and System Integration

PVDF-HFP is a fluorinated copolymer of vinylidene fluoride and hexafluoropropylene with a molecular weight of 254–293 K Daltons as measured by gel permeation chromatography (GPC) (Fig. 1).

Unlike many fluorinated polymers, for example, polytetrafluoroethylene (PTFE), PVDF-HFP is soluble in select solvents, rendering it suitable for coating applications. Everolimus and PVDF-HFP are dissolved in a solvent system designed for high solvency, good coating quality, and complete solvent removal. The ratio of everolimus to PVDF-HFP is 17/83 (w/w) in the coating formulation. After the primer coating is applied, the drug-polymer solution is conformally coated. The final coating contains 100 μg drug per cm^2 of stent surface area.

Similar to many fluorinated polymers, the PVDF-HFP has low surface tension[5] and its coating is lubricious and nonsticky, which minimizes unwanted adhesion to the balloon and between struts. A PBMA primer layer assures good adhesion of the PVDF-HFP coating to the stent. PBMA has been utilized in other blood-contacting permanent implants[6] and adheres well to the metal oxide layer on the stent surface via a combination of noncovalent hydrophobic, polar, and hydrogen bonding interactions. PBMA is partially miscible with PVDF-HFP. Hence, the adhesion of the PVDF-HFP drug reservoir to the primer layer is high as the PBMA and PVDF-HFP polymer chains are entangled at the interface. The thickness of the primer layer is only about 1 μm, which minimizes the coated strut dimensions and the polymer load on the stent.

Stent retention processes for XIENCE V drug-coated stent were developed to assure that stent security, system profile, and stent deliverability are equivalent to the ML VISION stent, while preserving the integrity of the coating. Stent dislodgement testing showed that the XIENCE V DES has equivalent stent retention to the ML VISION stent. Engineering studies

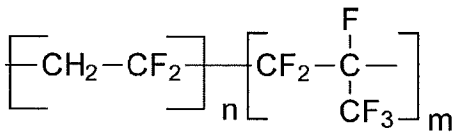

Figure 1. Chemical structure of PVDF-HFP.

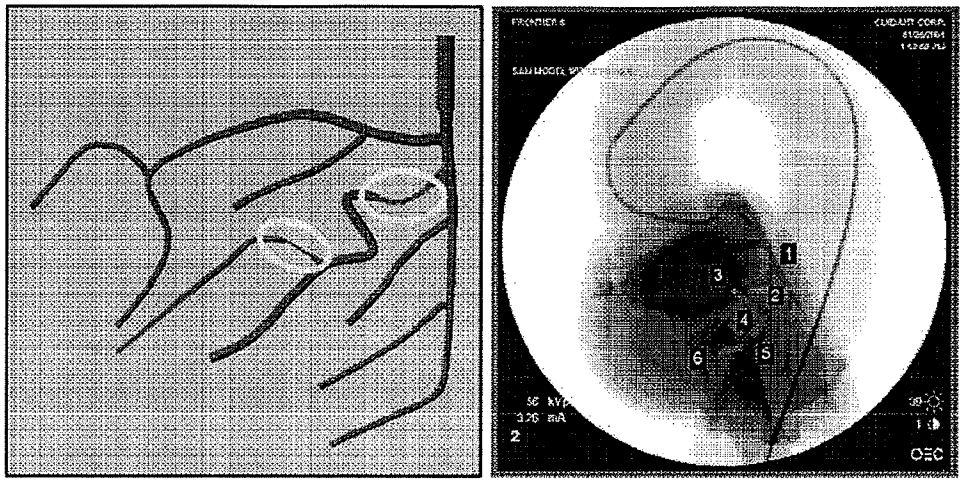

Figure 2. Design of synthetic vasculature model: Lumen of coronary artery (left panel) and an angiogram of the model (right panel). Numbers correspond to curves in track path of model.

have also demonstrated that the deliverability of the XIENCE V is equivalent to ML VISION in a synthetic vasculature model (Fig. 2). In this model, the stent delivery system was introduced by an experienced operator and advanced through the tortuosity until it could be advanced no further. A numerical score was assigned depending on how far the system progressed. Figure 3 summarizes data comparing XIENCE V with ML VISION. For all of the sizes tested in this model, the XIENCE V DES has demonstrated equivalent deliverability to ML VISION. XIENCE V also maintains a low system profile at 0.041″ (1.0 mm), which is comparable to the ML VISION stent profile of 0.040″ (1.0 mm) for 3.0 × 18 mm stent.

Hemocompatibility and Biocompatibility

Hemocompatibility (acute) and vascular biocompatibility (long-term) are of foremost importance in selecting a polymer as a drug-eluting reservoir. The characteristics determining the hemocompatibility and biocompatibility of an implanted biomaterial are many and complex and encompass more than just concepts

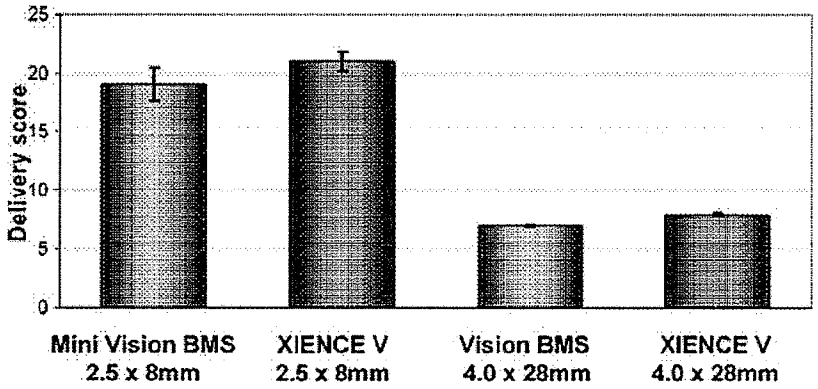

Figure 3. Illustration of synthetic test model and comparison data of XIENCE V DES deliverability with ML VISION bare metal stent (BMS) (n = 5). A higher score indicates better deliverability.

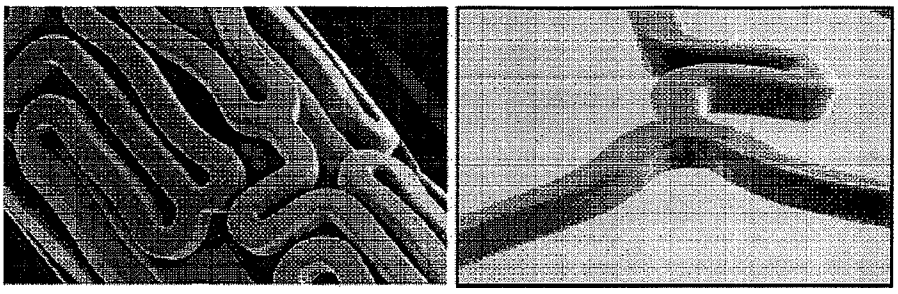

Figure 4. SEM images of the XIENCE V DES (scale = 150× in the left and 80× in the right).

of hydrophilicity and hydrophobicity. Factors such as surface chemistry, surface roughness, polymer purity, the presence of leachables and other impurities, in vivo stability of the polymer, and the interaction with blood-borne proteins all contribute to the blood and tissue compatibility of a polymer coating.

A critical first step in the interaction of a biomaterial with blood is the deposition of a protein-rich layer on the surface of the biomaterial. A high ratio of albumin to fibrinogen in this layer has been shown to greatly improve biocompatibility and hemocompatibility.[7–10] A series of studies[11–18] have shown that fluorine-rich surfaces are nonthrombogenic and biocompatibile. Such surfaces adsorb higher ratios of albumin to fibrinogen, resulting in significant reductions in platelet adhesion and activation.[11,12] Albumin adsorption passivates the fluorinated polymer surface, which contributes to thromboresistance, lack of complement activation, and the lower inflammation of fluorinated polymers in blood-contact applications in animal and in human.[13–16] Fluorination of polymer surfaces also increases the rate of reendothelialization of the vessel wall in animal models.[14] Due to their excellent hemocompatibility and biocompatibility, fluorinated polymers enjoy wide application in arterial prostheses, graft prostheses, drug-eluting leads of implantable cardioverter-defibrillators, hemodialysis membranes, vascular and neurovascular sutures, guidewire coatings, and in guiding catheters.[19] Preclinical studies performed at Abbott Vascular further confirm the hemocompatibility and biocompatibility of the XIENCE V PVDF-HFP polymer. These in vivo studies are discussed in a separate article in this supplement issue.

Polymer Properties. The PVDF-HFP polymer backbone is composed entirely of saturated carbon-carbon single bonds, which improves oxidative stability. These backbone carbon atoms are over 50%

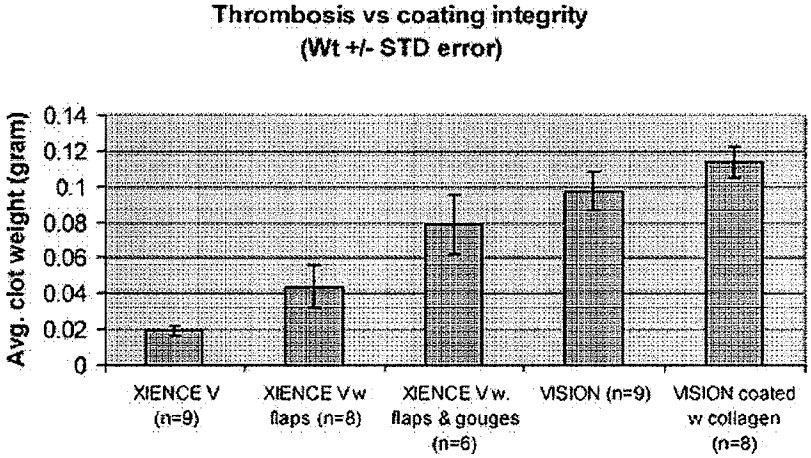

Figure 5. Relationship of thrombosis and coating integrity investigated in an ex vivo porcine femoral AV shunt model (3.0 × 18 mm stents).

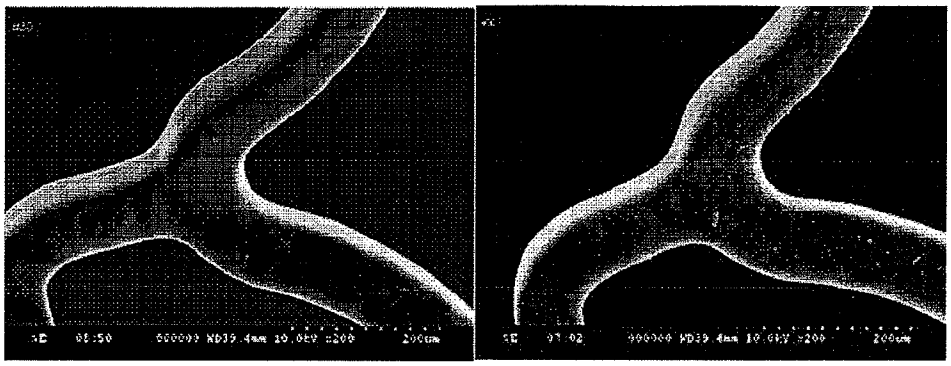

Figure 6. Representative SEM images of 10-year-accelerated fatigue test on overlapped XIENCE V DES with bending. The left image is prefatigue test and the right image is postfatigue test on nonoverlapped region. No change is observed.

fluorinated. The significance of fluorination is the high dissociation energy of the C-F bond that confers a high degree of chemical stability. Moreover, few carbon atoms with hydrogen substitution are adjacent to one another. This makes hydrogen radical abstraction to form carbon double bonds, a path to polymer degradation, unlikely. The absence of any reactive or enzymatically sensitive groups such as anhydride, ester, amide, ether, ketone, aldehyde, carbonate, or phosphate bonds makes the polymer resistant to hydrolytic, oxidative, or enzymatic cleavage. The lack of chemical reactivity of the polymer ensures no chemical interactions occur between PVDF-HFP and everolimus during the manufacturing process, on the shelf or in vivo. Similarly, this stable structure also ensures no chemical degradation takes place in vivo. Polymer chemical stability and the coating integrity were investigated in a swine model. In this study, everolimus-PVDF-HFP stents with a 200 $\mu g/cm^2$ drug dose were implanted for 3 months. After explantation, the polymer molecular weight was measured by gel permeation chromatography. For control stents stored at room temperature, the molecular weight was 287 K $\pm$ 2 K Daltons (n = 20), and for the explanted stents it was 284 K $\pm$ 13 K Daltons (n = 25). The difference was not statistically significant (P = 0.32), indicating the stability of the polymer in vivo.

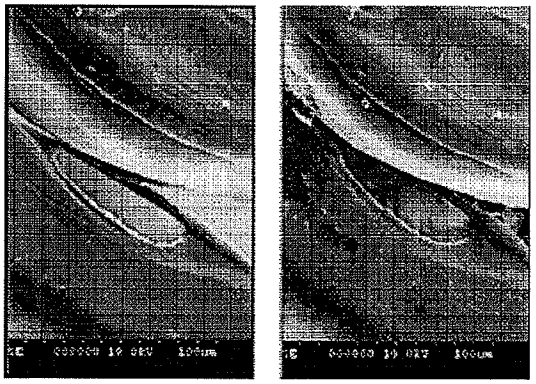

Figure 7. SEM images of 10-year-accelerated fatigue test on overlapped XIENCE V DES with bending. The left image shows the coating damage in the overlap region prefatigue test and the right images show the same damage region postfatigue test. No change is observed.

In this same study, the polymer mass was measured by ^{19}F nuclear magnetic resonance (NMR). After extracting the polymer off the stents, ^{19}F NMR spectroscopy allowed for selective quantification of the PVDF-HFP polymer. For controls stored at room temperature for 3 months, the mass recovery was 95.8 $\pm$ 3.0% (n = 5). For stents explanted after 3 months implantation, this recovery was 93.3 $\pm$ 2.3% (n = 5). The difference between the two groups was not statistically significant (P = 0.17), indicating the coating mass integrity was retained in vivo. These data demonstrate the stability of both molecular weight and mass for the coating and suggests that risk of inflammatory or other adverse biological responses caused by polymer degradation can be excluded.

For implantable materials, the purity of the polymer is essential for assuring a high level of biocompatibility. Total organic carbon extractable tests per United States Pharmacopeia (USP) on PVDF-HFP polymer have shown low levels of organic extractables at ppm levels. In addition, inductively coupled plasma mass

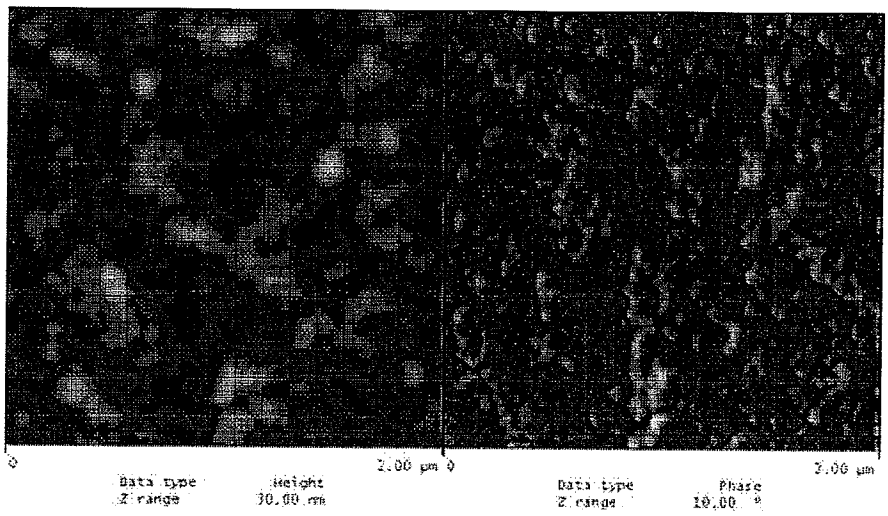

Figure 8. AFM topography image (left) and phase image (right) of XIENCE V coating on a 2 × 2 μm cross-section surface.

spectroscopy (ICP-MS) tests indicate that concentrations of heavy metals are at a sub-ppm level, thereby minimizing the risk of an inflammatory response or other adverse biological responses due to impurities.

Coating Integrity. For DES, the integrity of the coating is another important factor that can affect the hemocompatibility and biocompatibility of the system. It can also influence the uniformity of the drug delivery to the tissue. Suboptimal coating integrity may lead to coating particles being shed during delivery and expansion of the stents. Significant sections of missing coating expose the metal stent to the tissue and reduce the local drug delivery, thereby reducing the benefit of the drug. Coating particles and fragments released into the vessel wall may also elicit an inflammatory response at the site of implantation. Maintaining the integrity of the coating was a key design criterion for the XIENCE V DES. The selection of a semicrystalline, but flexible, fluorinated polymer, the design of the primer and drug reservoir construct, and the

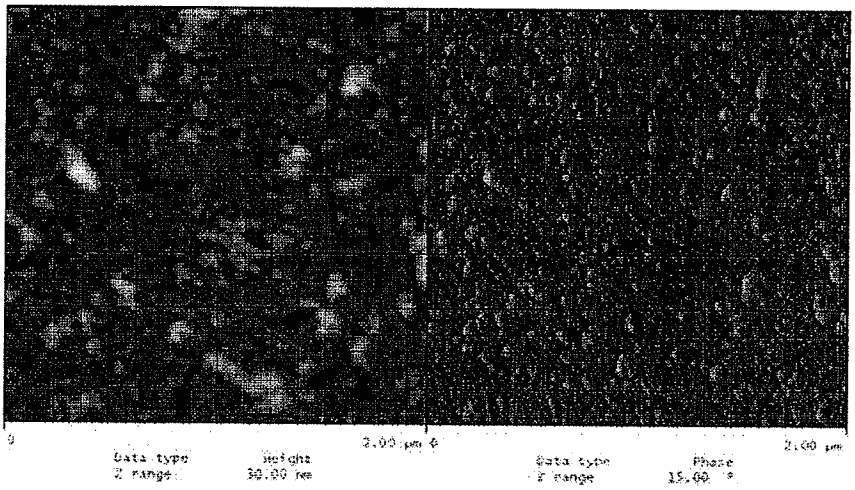

Figure 9. AFM topography image (left) and phase image (right) of polymer-only coating on a 2 × 2 μm cross-section surface.

Table 1. Thermal Properties of XIENCE V Coatings Relative to PVDF-HFP/PBMA Coatings with No Drug

ΔH_f, % of Polymer-Only Stent (n = 6)	Melting Point % of Polymer-Only Stent (n = 6)
104 ± 8.6	99.0 ± 0.91

Table 2. Thermal Properties of XIENCE V Coatings after Aging at Ambient Conditions

Age	Drug Transition Temp., % of XIENCE V at T_0 (n = 6)	ΔH_f, % of XIENCE V at T_0 (n = 6)
20 months	101 ± 1.9	98.6 ± 8.2
28 months	100 ± 1.8	92.4 ± 9.4

optimization of solvent-drug-polymer formulation resulted in exceptional coating integrity for the XIENCE V DES. Scanning electron microscope (SEM) analysis and bench testing confirm a smooth coating surface with minimal to no structural defects such as webbing, bridging, and strut-to-strut sticking (see Characterization section). Representative images of the XIENCE V coating as produced and after expansion are shown in Figure 4.

An acute ex vivo porcine femoral arteriovenous (AV) shunt model was employed to evaluate the influence of the coating defects on the hemocompatibility of XIENCE V stent. The model assessed thrombus formation and platelet deposition on XIENCE V stents as well as two groups of XIENCE V stents with intentionally modified coating surface. One group of XIENCE V stents were manufactured with luminal flaps and end ring defects. The second group had these same defects, but in addition, were intentionally scratched and the surface roughened with gauges. Controls were bare metal ML VISION and ML VISION coated with type I collagen. The XIENCE V DES exhibited minimal thrombus formation on the stent struts. Stents with more coating defects were more thrombogenic. ML VISION stents coated with hydrophilic collagen exhibited the highest thrombus formation. These study results demonstrated that the smooth and conformal fluorinated polymer coating surface on XIENCE V DES improves the thromboresistance (Fig. 5).

XIENCE V Coating Physical Characterization

Due to its glass transition temperature (T_g) of approximately –29 °C, the PVDF-HFP polymer is elastic with an ultimate elongation of over 600%. Such elasticity is essential for the coating to withstand stent deployment and postdilation without coating cracking or delaminating. PVDF-HFP is a partially crystalline polymer with a relatively low degree of crystallinity. Due to its low T_g and crystallinity, PVDF-HFP is tough and fatigue resistant. This allows the XIENCE V coating to withstand stent crimping forces and the abrasion/shearing forces encountered when delivering the stent to a challenging lesion with minimal coating damage. Engineering tests have demonstrated that the coating integrity is maintained after in vitro simulated delivery tests to the maximum labeled expansion of the stent, and after 10 years of accelerated radial fatigue testing (Figs. 6 and 7). In Figure 7, preexisting coating defects did not propagate after 10 years of

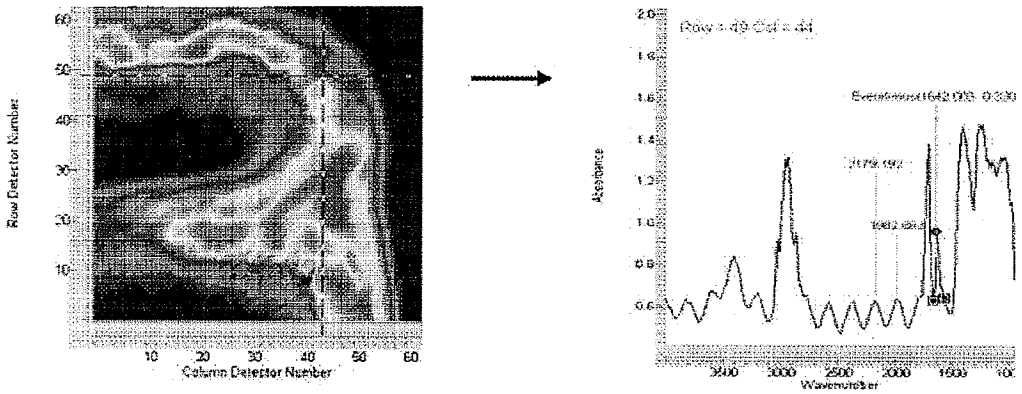

Figure 10. Representative FTIR drug distribution image on XIENCE V DES and FTIR spectrum showing interference pattern.

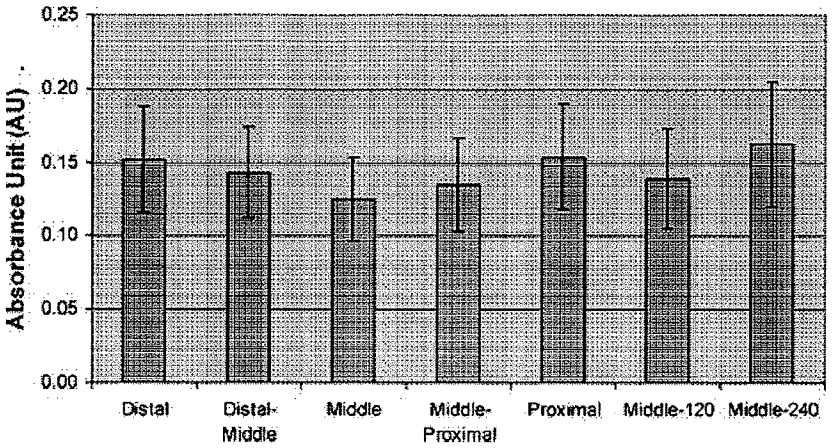

Figure 11. Drug distribution longitudinally (distal, distal-middle, middle, proximal, proximal-middle, and proximal) and circumferentially (120° and 240° rotation from middle) by imaging FTIR measurement (n = 30 spots per location).

accelerated fatigue testing. Chronic coating durability tests performed after tracking the preconditioned stents through tortuous paths, and deployment in synthetic arteries, show particulate counts comparable to that of the control blank synthetic artery, thereby indicating that the XIENCE V coating is durable.

Physical characterization studies indicate that the everolimus is uniformly dispersed in the polymer matrix at a nanometer scale as evidenced by atomic force microscopy (AFM) studies. Figure 8 shows a cross-section of the drug-polymer reservoir of a XIENCE V coating at a 2 × 2 μm scan size. At this magnification, the coating displays a high level of homogeneity over the entire cross-section. A fine, granular texture in the range of 70 nm is visible. A similar granular texture was also visible on cross-sectional analysis of a polymer-only stent (Fig. 9). The granular texture does not appear to be drug related and is believed to be due to the polymer. Consequently, no signs of phase separation between drug and polymer were detectable within the nanometer scale resolution of the method.

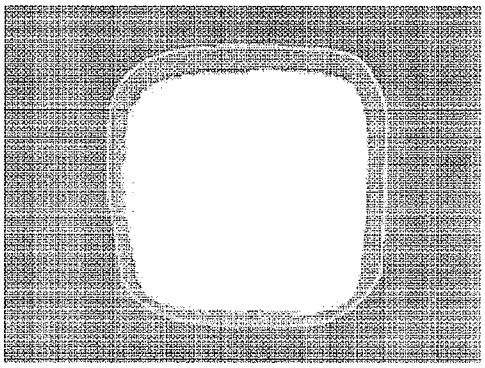

Figure 12. Representative cross-sectional SEM image of a XIENCE V stent strut with coating.

Thermal analysis was performed on XIENCE V stent coatings by differential scanning calorimetry (DSC). The thermograms show a melting endotherm due to the crystalline phase of the PVDF-HFP and a thermal transition due to the drug. A summary of these data and those from stent coatings with no drug is given in Table 1.

The presence of the drug in the polymer does not appreciably alter the PVDF-HFP degree of crystallinity or melting point. This indicates that coating properties that depend on polymer crystallinity, such as hardness and ultimate tensile strength, are less affected. Bulk everolimus exhibits a glass transition (T_g) at 87 °C. In the XIENCE V coating, the everolimus T_g is shifted down to 76 °C, with no transition evident at 87 °C. These data support the concept that no bulk everolimus is present in the coating. DSC analysis also shows that the melting enthalpy of the PVDF-HFP and the thermal transition temperature of everolimus do not change over time, indicating that the drug coating is at thermodynamic equilibrium after manufacture and sterilization (Table 2).

The distribution of the drug longitudinally and circumferentially in the XIENCE V coating was investigated by focal plane array imaging Fourier transform infrared spectroscopy (FTIR) in reflectance mode.

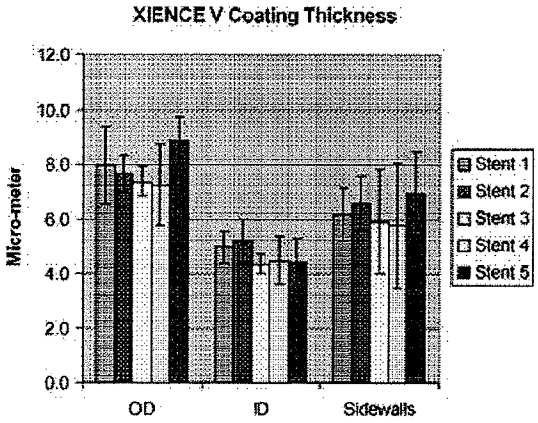

Figure 13. XIENCE V coating thickness data by cross-sectioning followed by SEM measurement. Fifteen struts per stent were sectioned for this analysis.

Figure 10 shows the XIENCE V drug distribution via this imaging technique where the spectra were gathered at an infrared wavelength specific to the drug. The relative drug concentration at each location is equivalent to the optical absorbance of this drug peak and can be used to quantify the relative drug distribution as a function of position. The color intensity corresponds to the total amount of drug integrated over the coating thickness at each point.

Three stents were used in this test. Seven stent positions covering the length and circumference were analyzed. At each location, absorption spectra for 10 randomly chosen points were obtained and averaged for the drug absorbance. Figure 11 shows the drug density (in infrared absorbance units) as a function of location on the stent. These results demonstrate a uniform drug distribution over the abluminal stent surface.

The coating thickness was measured by cross-sectioning the coated stent, followed by SEM analysis. The coated stents were first coated with a protective layer. The stent was then embedded in a resinous medium, sawed to make rough cross-sections and then polished. SEM images of each strut in cross-section were obtained (Fig. 12).

Care was taken to assure that the stent axis was orthogonal to the sectioning plane. The average coating thicknesses over the abluminal, luminal, and sidewall surfaces were measured with image analysis software (Fig. 13). The average coating thickness on the abluminal (OD) surface by this method was 7.8 ± 0.7 μm, and on the luminal surface (ID), 4.7 ± 0.4 μm.

Considering the drug characteristics in the polymer coating, the drug release mechanism is based primarily on the molecular diffusion of the drug through the polymer. This release mechanism results in a system with inherently high reproducibility for the drug release rate. The uniform drug distribution at both the micro- and macroscales also enables the tight control of the drug release rate across different manufacturing lines, from lot-to-lot and across multiple stent sizes to ensure precise delivery of the drug dose to patients (Fig. 14).

Conclusion

XIENCE V coating system is the result of research on multiple polymers, combined with extensive coating

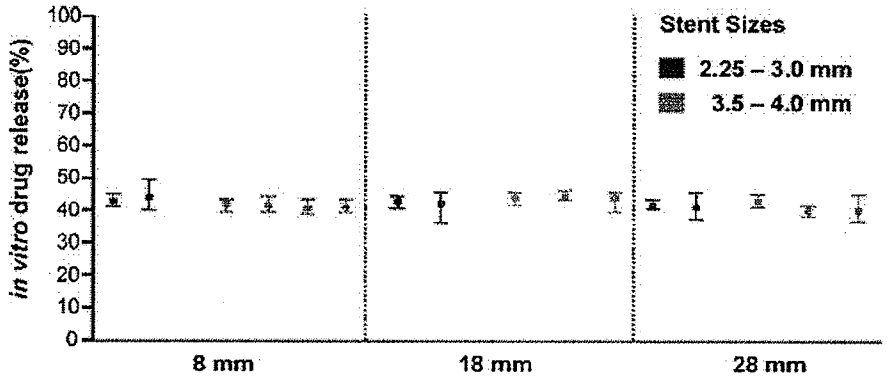

Figure 14. In vitro percent drug release of Xience V DES demonstrates consistent drug release rate on different stent sizes, lot-to-lot, and within the lot. Stents from the same lot were coated on different coating equipment.

formulation and design iteration. A fluorinated copolymer drug reservoir coating was chosen based on the history of use of fluorinated polymers in a blood-contacting environment, the ability to control the drug release, and mechanical properties amenable to stent expansion and use. Bench characterization and preclinical evaluation results, as discussed in a separate article in this supplement issue, demonstrate that XIENCE V is highly deliverable, and the coating is biocompatible and possesses exceptional coating integrity. Clinical data from the SPIRIT series of human clinical trials[1,2,20] support the safety and effectiveness of the XIENCE V, a second-generation DES.

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